FREQUENCY RESPONSE OF THE HUMAN SEMICIRCULAR CANALS

II. NYSTAGMUS PHASE SHIFT AS A MEASURE OF NONLINEARITIES

BY

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JOINT REPORT

UNITED STATES NAVAL SCHOOL OF AVIATION MEDICINE

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SUMMARY PAGE

THE PROBLEM

A major source of potential operational problems for man-in-space programs is the high-level angular accelerations which may occur during either normal or abnormal flight profiles. The system transfer function concept has been applied to the description of the steady-state response of the oculovestibular system to sinusoidal angular acceleration stimuli, and the feasibility of experimental identification and quantification of the transfer functions has been demonstrated. The existence of nonlinearities may, however, be crucial for the interpretation and integration of data obtained in a wide variety of situations ranging from low-level, near physiological test applications to high-level aerospace environments.

FINDINGS

The lag of the steady-state nystagmic eye velocity response behind a sinusoidal angular acceleration stimulus is used to quantify the transfer function for the skull-acceleration-to-cupula-endolymph-motion transducer system. The experimental measure of this lag was derived from corneoretinal potential recordings and expressed in standard electrical degree form to describe the phase difference angle between stimulus and response as a function of both frequency and magnitude of the stimulus. The effects of sinusoidal angular acceleration stimuli ranging in frequency from 0.02 cps to 0.20 cps and in peak acceleration level from 10 deg/sec^2 to 80 deg/sec^2 were evaluated for four subjects.

The nystagmic phase shift data demonstrate independence of stimulus magnitude only at the upper frequencies; at the lower frequencies, phase lag varied inversely with the peak acceleration level indicating nonlinear operation.

The experimental and theoretical considerations involved in using the nystagmus transition technique to study these nonlinearities are discussed and an illustrative application of their quantification is presented.

INTRODUCTION

In a previous report (4) the system transfer function concept was developed to describe the steady-state response of the oculovestibular system to periodic angular acceleration of sinusoidal form. With this concept it was possible to define three related system transfer functions: $K_0 G_0$ (s), $K_1 G_1$ (s), and $K_2 G_2$ (s). The first described the over-all transduction process whereby angular acceleration of the skull produces nystagmic eye motions, and is equivalent to the product of K_1 G_1 (s) which described that element of the system responsible for the transduction of the skull accelerations to cupula-endolymph motions within the semicircular canals and K_2 G_2 (s) which was concerned with the conversion of these cupula-endolymph motions to nystagmic eye movements via the central nervous system integrative pathways.

It was then demonstrated that the separate identification and quantification of each transfer function could be approached under actual experimental conditions by relating objective measures of eye velocity during the slow component of nystagmus to periodic angular acceleration stimuli of variable frequency. Specifically, the lag of the steady-state nystagmic eye velocity response behind a sinusoidal driving stimulus was used to quantify K_1 G_1 (s) in the form used by Groen (2) to represent Steinhausen's (6) equation of motion for the torsion pendulum behavior of the canal mechanism. This lag, experimentally measured by utilizing a transition technique which is relatively independent of the perturbations in eye displacement or velocity commonly encountered during the recording of nystagmus, was expressed in standard electrical degree form to describe the frequency dependent phase difference between the stimulus and the response. For quantification of the K_2 G_2 (s) transfer function, the actual peak velocity which occurred during the slow component of the steady-state nystagmus response to the same sinusoidal stimuli served as the basic measure.

This latter function can be expressed alternatively in terms of eye displacement by considering the integrative action of the CNS pathways which is periodically reset during each fast component of nystagmus. However, in the experimental situation, variations in the instantaneous eye displacement baseline make its quantification much more difficult than the related eye velocity measure.

With use of the frequency response representation of these system transfer functions, the phase difference between the input stimulus and output response of each component of the system was defined (4) in vector nomenclature form as follows:

$\angle\, \phi_0\, (j\omega)$ = experimentally measured phase difference between the acceleration stimulus and the nystagmus response

$\angle G_0(j\omega)$ = theoretical phase difference between the acceleration stimulus and the nystagmus response

$\angle G_1(j\omega)$ = theoretical phase difference between the acceleration stimulus and the cupula displacement response

$\angle G_2(j\omega)$ = theoretical phase difference between the cupula displacement stimulus and the nystagmus response

where

$$\angle G_0(j\omega) = \angle G_1(j\omega) + \angle G_2(j\omega).$$

Because the response speed capability of the oculomotor mechanisms far exceeds that of the cupula-endolymph system, $\angle G_2(j\omega)$ was considered to be negligible compared to $\angle G_1(j\omega)$ over the limited range of frequencies selected to stimulate the canals. This assumption implies that the over-all phase lag of the nystagmus response behind the acceleration stimulus should be attributable only to the characteristics of the canals, i.e., that the eye velocity is essentially in phase with the cupula displacement and that $\angle G_0(j\omega)$ is very nearly equal to $\angle G_1(j\omega)$. Then, if linearity of response is to exist over a limited range of operation, $\angle G_0(j\omega)$ should be a function of the frequency, but not of the magnitude, of the stimulus. However, particular attention was always given to defining separately $\angle \phi_0(j\omega)$, an experimentally measured phase angle, and $\angle G_0(j\omega)$, a theoretical description of the same phase angle. This distinction permits the validity of any synthesized representation of $\angle G_0(j\omega)$ to be evaluated by direct comparison with $\angle \phi_0(j\omega)$.

In an associated report (5), these theoretical and experimental approaches were combined to study steady-state vestibular response to variable frequency sinusoidal angular acceleration stimuli of the form used by such investigators as Wendt (7), von Békésy (1), and Hennebert (3). A constant peak acceleration level of 40 deg/sec^2 was used to evaluate the nystagmus phase shift occuring over the 0.02 to 0.2 cps frequency spectrum. The selection of this relatively high stimulus level, based on a hypothetical aerospace environment, enabled us to show that nonlinearities would limit the extent to which the actual $\angle \phi_0(j\omega)$ data could be aproximated by the synthesized $\angle G_1(j\omega)$ component of the overall labyrinth transfer function. In particular, for one subject, it was shown that calculated values for the parameters of the canals could not be interpreted in linear fashion since the $\angle \phi_0(j\omega)$ data at the 0.02 cps stimulus frequency was not independent of the peak magnitude of the acceleration.

Since the existence of nonlinearities may be crucial for the interpretation and integration of findings obtained in a wide variety of situations ranging from low-level, near-physiological, test applications to high-level aerospace environments, an exploratory search of the potential limits was indicated. The present report presents data on nystagmus phase shift response to sinusoidal angular accelerations of variable frequency and variable peak magnitude. It also serves to demonstrate the effectiveness of the nystagmus transition technique as a measure of the linearity of the steady state oculovestibular response.

APPARATUS

Sinusoidal angular accelerations of varying frequency and peak acceleration level were used as stimuli in the present experiment. To produce these stimulus patterns, the capsule of the Human Disorientation Device (HDD) was rotated about its vertical axis with the output of a low-frequency generator (Hewlett-Packard Model 202A) serving as the velocity input signal to the power servomechanism drive system of the device. The selected combinations of frequency and peak acceleration levels are given in Table I along with the phase shift data of the HDD proper at each of the stimulus frequencies. As described previously (5), these data account for the frequency dependent lag characteristics of the HDD and are used in the reduction of the actual nystagmus phase shift data.

TABLE I

Experimental Stimuli Combinations

Cyclic Frequency (cps)	Cyclic Period (sec)	Peak Acceleration (deg/sec²)	HDD Phase Shift (deg)
0.02	50.0	10, 20, 30, 40	0
0.04	25.0	20, 30, 40,60, 80	1.6
0.08	12.5	30, 40, 60, 80	3.4
0.20	5.0	40, 60, 80	7.0

RESPONSE

Recordings of nystagmic displacements of the eye in the horizontal direction were obtained by the corneo-retinal potential method. The displacement signal derived from silver electrodes placed at the outer canthi was amplified by a differential input preamplifier (Taber Model 202G-4) and transmitted via slip-rings to one channel of a direct-writing recorder (Sanborn Model 358) after filtering to minimize muscle-potential artifacts.

PROCEDURE

EXPERIMENTAL

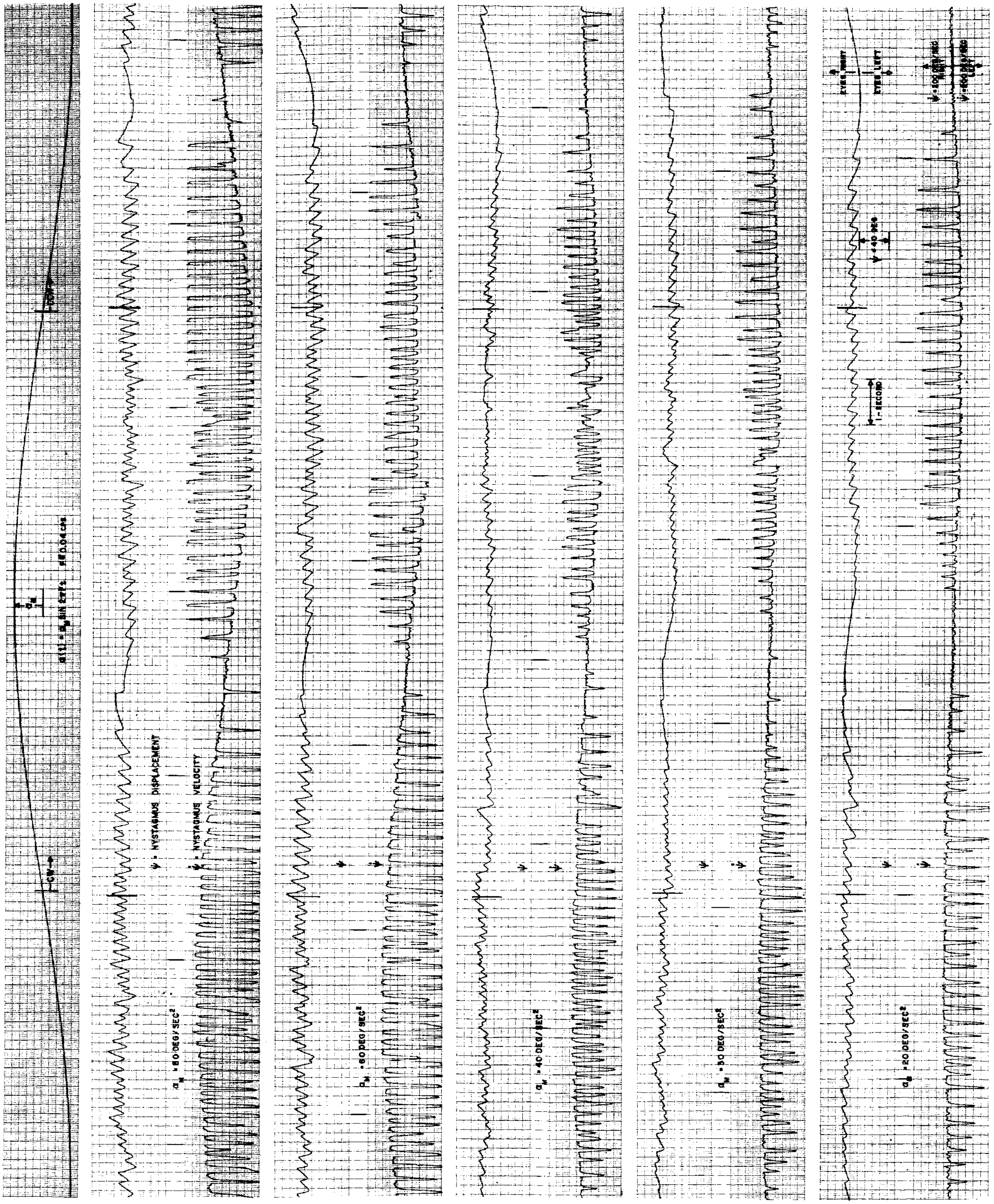
Four volunteer male subjects, 22 and 23 years of age, participated in the present study. All were healthy medical students, on active duty in Naval reserve officer status, who had no known defects of hearing or equilibrium. Instruction was given to each of them regarding the purpose of the experiment, the procedure for recording nystagmus, and the nature of the rotational stimuli he would experience.

All runs were made in total darkness while the subject, with his eyes open, was seated securely, his head fixed on the vertical axis of rotation. At the beginning of every experimental session, the nystagmus recording circuits were calibrated by having the subject track a visual target with an over-all excursion of 40 degrees of arc. During these sessions, he was exposed to oscillation at two test frequencies, each being presented at several levels of peak angular acceleration (see Table I). Since a previous study (5) had shown no significant effects due to the order of presentation of test frequencies, the subject was exposed to 0.20 cps and 0.02 cps on his first session and 0.08 cps and 0.04 cps on his second session. Duplicate runs were made subsequently. For each test frequency, the level of peak angular acceleration was always increased from the lowest value to the highest, since immediate exposure to high levels of acceleration stimuli tended to create anxiety which influenced adversely the nystamus recording quality.

The subjects were exposed to each test frequency-peak acceleration stimulus combination for sixty seconds before the actual recording began to eliminate potentially significant transient effects and to ensure the establishment of a steady-state mode of response. Theoretical computations using Groen's labyrinth constants for a typical subject indicated that forty-six seconds would be adequate, but allowance of a margin for individual differences led to the selection of a 100-second pre-recording oscillation interval in a previous study (5); those experiments further indicated the feasibility of decreasing this allowance and hence the reduction of the preliminary oscillation period to sixty seconds.

ANALYTICAL

Recordings of nystagmus responses produced by the sinusoidal angular acceleration stimulus profile for a representative test frequency and increasing peak angular acceleration level are shown for a single subject in Figure I. The waveform of the 0.04 cps stimulus is shown at the top with the onset of clockwise or counterclockwise acceleration indicated by vertical bars on both this record and the nystagmus data below. For each of the five acceleration levels (80, 60, 40, 30, and 20 deg/sec^2) the top trace represents instantaneous eye displacement, and the trace immediately beneath, the related eye velocity as obtained by electrical differentiation of the displacement data. The characteristic saw-tooth pattern of nystagmus is readily apparent as is the transition of nystagmus direction in response to the changing sign of the acceleration stimulus. The sinusoidal variation of eye velocity is also displayed in vivid fashion by these recordings.



Corneo-retinal potential recordings of nystagmic eye displacements and velocities during vestibular stimulation by sinusoidal angular accelerations of fixed frequency and increasing peak acceleration level (see text).

FIGURE 1

As described previously (5), the interval between the onset of angular acceleration in one direction and the instant at which the following transition of nystagmus direction occurred served as the basic measure of nystagmus phase shift. This interval was compared to the cyclic period of the stimulus and expressed in conventional electrical degree nomenclature where one cycle of the stimulus represents 360 degrees. Since the control signal was used as a reference for the determination of the time of acceleration onset, corrections for the lag characteristics of the HDD were made by subtracting the related phase shift data of Table I from the nystagmus phase shift measurements.

RESULTS AND DISCUSSION

The data derived from the four test subjects for the various combinations of frequency and peak acceleration levels used as the sinusoidal angular acceleration stimuli are summarized in Table II. These data are also shown in Figure 2 as a plot of nystagmus phase lag in electrical degrees versus the peak magnitude of the acceleration stimuli for various cyclic frequencies. The general pattern of these data is that the phase shifts at the higher stimulus frequencies (0.20 and 0.08 cps) are relatively independent of the magnitude of the peak acceleration of the stmulus, whereas

TABLE II

Mean Phase Shift Angle* as a Function of Peak Acceleration at Selected Stimulus Frequencies for Each of Four Subjects.

Frequency (cps)	Peak Acceleration (deg/sec²) 10	20	30	40	60	80
SUBJECT JEW						
0.02	56.15	49.32	43.35	39.3	----	----
0.04	----	61.17	58.15	56.2	52.37	49.75
0.08	----	----	73.67	71.12	70.57	67.97
0.20	----	----	----	84.87	84.27	83.25
SUBJECT RBH						
0.02	56.1	55.5	51.1	49.18	----	----
0.04	----	66.48	64.38	64.58	63.88	63.32
0.08	----	----	74.62	73.85	75.08	72.98
0.20	----	----	----	83.08	82.92	82.38
SUBJECT LWH						
0.02	61.25	57.0	52.52	49.15	----	----
0.04	----	70.27	69.35	68.12	65.02	62.92
0.08	----	----	77.77	76.25	76.05	76.1
0.20	----	----	----	80.5	81.97	83.17
SUBJECT DBG						
0.02	49.82	48.17	45.17	41.67	----	----
0.04	----	64.3	61.07	60.95	57.52	56.62
0.08	----	----	72.07	72.57	72.85	73.35
0.20	----	----	----	79.45	78.5	79.9

*Each datum representing the mean of eight measures, combining two directions and two orders of presentation of the sinusoidal acceleration stimuli, is expressed in electrical degrees and has been corrected for HDD machine phase shift at each frequency (see text).

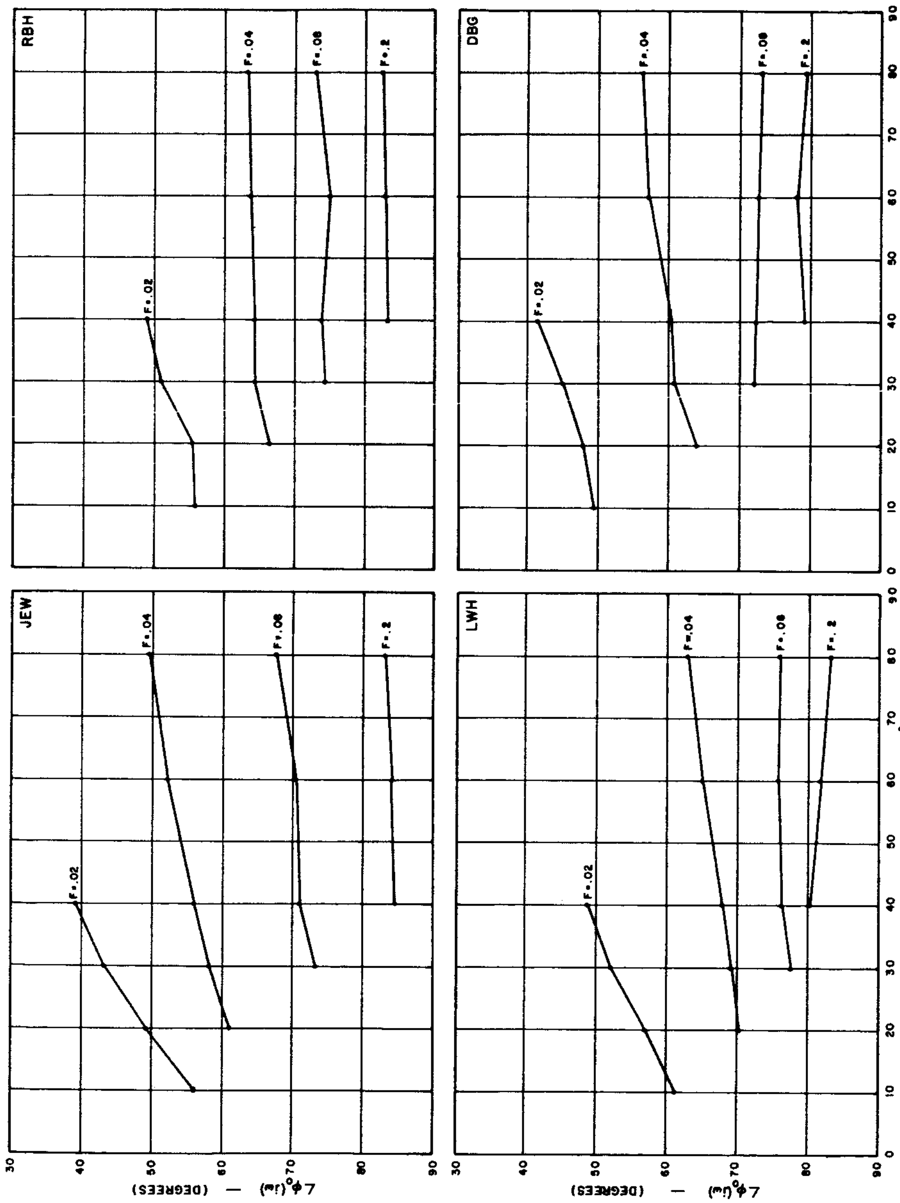
at the lower test frequencies (0.02 and 0.04 cps) the phase shift consistently varies inversely with the peak acceleration level. It may also be noted that the slope of the 0.02 cps phase shift curve for a given subject is aways greater than that of the corresponding curve for his response to the 0.04 cps stimulus.

Significant intersubject differences can also be observed by comparing the Figure 2 data of RBH with those of JEW. The phase shift of RBH is relatively independent of the peak acceleration at all frequencies except for the changes which occur at the 30 and 40 deg/sec^2 levels of the 0.02 cps stimulus. However, the data of JEW clearly indicate that his phase shift cannot be considered as independent of the stimulus level even at the higher frequencies and that at the 0.02 cps frequency, the slope of his curve is much greater than that of RBH.

The consequences of these variations in phase shift, which occur when the magnitude of the acceleration is changed, for the apparent frequency response characteristics of the oculovestibular system are illustrated in Figure 3 which is a plot of the Table II nystagmus phase shift data versus the cyclic frequency of the sinusoidal stimulus for various acceleration levels. With this conventional frequency response representation of phase shift data, a rough estimate of the damping present in the system can be made by observing the relative slope of the phase curve; i.e., a system with little damping will have a greater rate of change in phase with frequency than a system that is heavily damped. Thus, if one were to plot only the nystagmus phase shifts which occurred at the highest acceleration levels used for the 0.02 and 0.04 cps stimulus frequencies, the system would appear to be much less damped than if the same curve were plotted using the lower acceleration levels. Incidental to this discussion, it should be noted from Figure 3 that the undamped characteristic cyclic frequency (the frequency at which $\phi_0 = 90$ degrees) for these four subjects is well above Groen's value of 0.159 cps (2) which was derived from measures of the subjective sensation of turning.

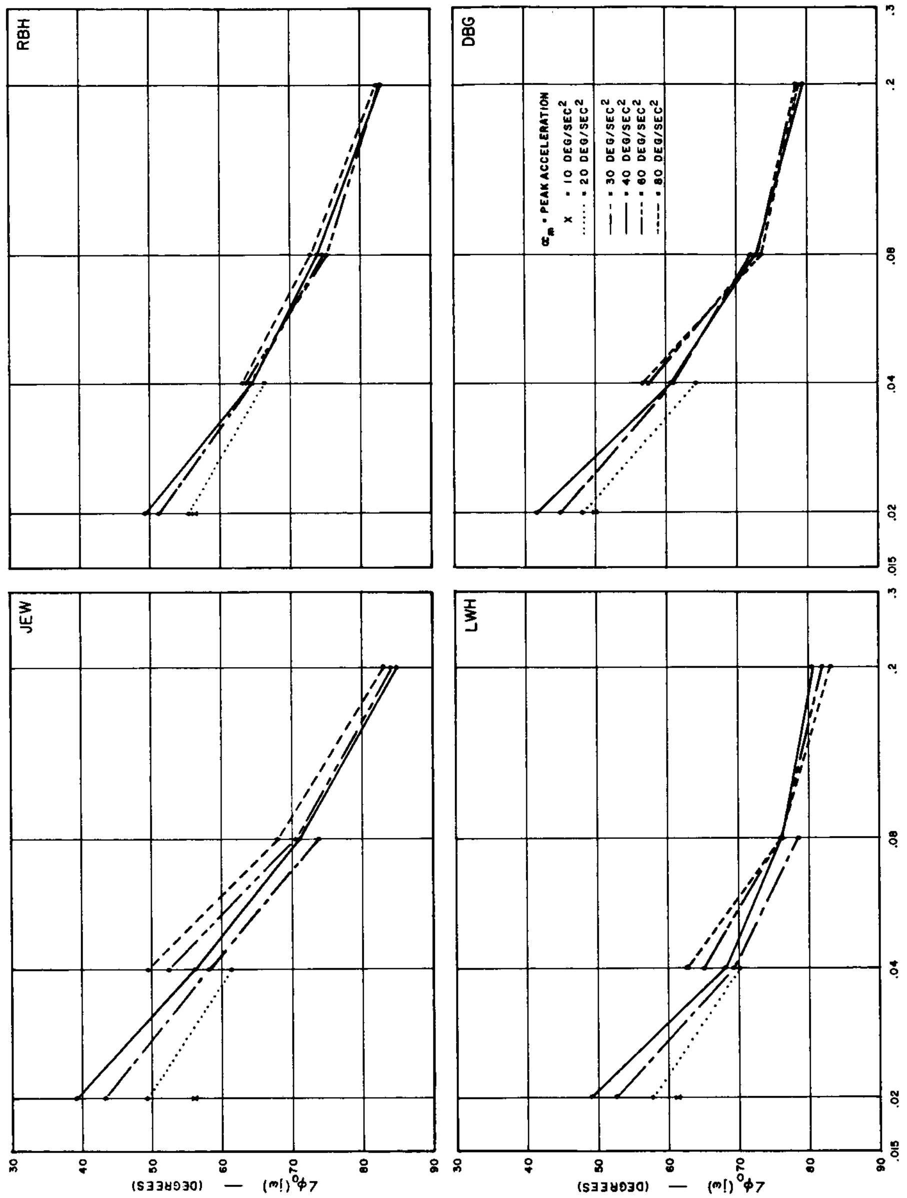
In the interpretation of these data, it is pertinent to review some of the factors involved in Steinhausen's concept that the cupula-endolymph structure acts as a heavily damped torsion pendulum and thus can be described by a linear differential equation of second-order. For the behavior of a physical or biological mechanism to be fully described by such a mathematical expression, each of its components must possess all of the characteristics demanded by this general class of nth order linear integrodifferential equations. Since the independent variables of these equations must be of first-order with constant, time-invariant coefficients, it is necessary that the analogous physical or biological mechanism be composed of linear elements, i.e., elements whose input-output characteristics are constant and can be related by a straight line.

With such a linear system, one of the most prominent operating characteristics is that the time-amplitude contour of the output response produced by a given time-varying excitation function will be independent of the amplitude scale factor associated with the magnitude of the input signal. When the level of a given excitation function is multiplied by a constant, the waveform of the output response will be unaltered except for



Nystagmus phase shift data from four subjects plotted versus the peak acceleration level of the sinusoidal angular acceleration stimulus for various cyclic frequencies.

FIGURE 2



Nystagmus phase shift data from four subjects plotted versus the cyclic frequency of the sinusoidal angular acceleration stimulus for various peak acceleration levels.

FIGURE 3

modification of its amplitude level by the same constant. For example, for step function input signals the time required for the output to reach a predetermined percentage of its final value will be fixed and independent of the magnitude of the driving signal; for sinusoidal excitation of fixed frequency the phase difference between the input stimulus and the output response will be completely independent of the peak magnitude of the excitation. These operating characteristics are directly related to the fundamental concept that the output response of a linear system cannot contain any frequency components which are not present in the spectrum of the applied excitation signal.

Although a physical or biological mechanism may satisfy these performance criteria over a given stimulus range, a level of input excitation can usually be found where operation in the linear mode is not feasible. Limitations inherent to the physical nature of the system components can result in saturation or overloading effects which will limit the stimulus range in which the constants can be considered as invariant and independent of the variables describing its performance. Similarly, physical or biological aging processes will limit the interval whereby the characteristics of the system or device can be considered as time-invariant. For the complete study of these nonlinearities interest must be directed both toward describing the amplitude-time characteristics of the deviations from linear operation and toward determining if exposure to this state produces any changes within the system. For biological mechanisms the extent and time course of any changes effected on their characteristics will determine the consequences, whether merely adaptive or possibly deleterious, of operation in the nonlinear mode.

For the specific case of exposure of the human semicircular canals to high-level stimuli which may produce nonlinear operation, consider Groen's equation of motion for the cupula-endolymph system which is:

$$\ddot{\xi} + \frac{\Pi}{\Theta}\dot{\xi} + \frac{\Delta}{\Theta}\xi = \alpha\,(t)\,,\ \xi \equiv \xi\,(t) \qquad (1a)$$

where $\xi,\ \dot{\xi},\ \ddot{\xi}$ = angular displacement, velocity, and acceleration of cupula

$\Theta,\ \Pi,\ \Delta$ = rotational mass, damping, and stiffness of cupula mechanism

$\alpha\,(t)$ = angular acceleration of skull

or in terms of the ζ and ω_n performance parameters utilized in the control and servomechanisms areas

$$\ddot{\xi} + 2\zeta\,\omega_n\,\dot{\xi} + \omega_n^2\,\xi = \alpha\,(t) \qquad (1b)$$

where

$$\zeta = \frac{\Pi}{2\,(\Delta\,\Theta)^{1/2}} \quad = \text{damping ratio}$$

and

$$\omega_n = \left(\frac{\Delta}{\Theta}\right)^{1/2} \quad = \text{undamped characteristic angular frequency.}$$

Since these expressions belong to the general class of *n*th order linear integrodifferential equations, it is necessary that the coefficients of the independent variables, i.e., the mechanical characteristics of the system being described, be constants for a given level of the input stimulus. Thus the equivalent moment of inertia, Θ, the rotational resistance or viscous damping, Π, and the rotational stiffness, Δ, parameters must be fixed, time-invariant quantities which are independent of the cupula motions if the canals are to be assumed to be operating in a linear mode.

When a sinusoidal angular acceleration stimulus of the form

$$\alpha\ (t) = \alpha_m \sin \omega t \tag{2}$$

where

$$\alpha_m = \text{peak magnitude of } \alpha\ (t)\ ,$$

$$\omega = 2\pi F = \text{angular frequency,}$$

and

$$F = \text{cyclic frequency}$$

is applied, the theoretical behavior of the canals will be such that the cupula motion will lag the stimulus by the angle defined as

$$\angle\ G_1\ (j\omega) = -\ \text{arc tan}\ \frac{2\zeta \dfrac{\omega}{\omega_n}}{1 - \dfrac{\omega^2}{\omega_n{}^2}} \tag{3a}$$

or as

$$\angle\ G_1\ (j\omega) = -\ \text{arc tan}\ \frac{\dfrac{\Pi}{\Theta}\ \omega}{\dfrac{\Delta}{\Theta} - \omega^2} \tag{3b}$$

By relating the actual measured nystagmus phase shift data $\angle \phi_0 (j\omega)$ to $\angle\ G_1 (j\omega)$, as was described in (4) and (5), it can be seen that the phase shift should be independent of the magnitude of the stimulus and be of fixed value for a given angular frequency if the system is operating in the linear mode.

When the data depicted in Figure 2 are again examined, it can be seen that these particular criteria for operation in the linear mode are approximated at the higher stimulus frequencies but certainly not at the lower frequencies. Since the theoretical phase lag produced by stimulus frequencies much lower than the undamped characteristic angular frequency ω_n of the canals can be closely approximated by

$$\angle\ G_1 (j\omega) \Big|_{\omega << \omega_n} \cong -\ \text{arc tan}\ \frac{\Pi}{\Delta}\ \omega \tag{4a}$$

or

$$\angle\ G_1 (j\omega) \Big|_{\omega << \omega_n} \cong -\ \text{arc tan}\ 2\zeta \frac{\omega}{\omega_n}\ , \tag{4b}$$

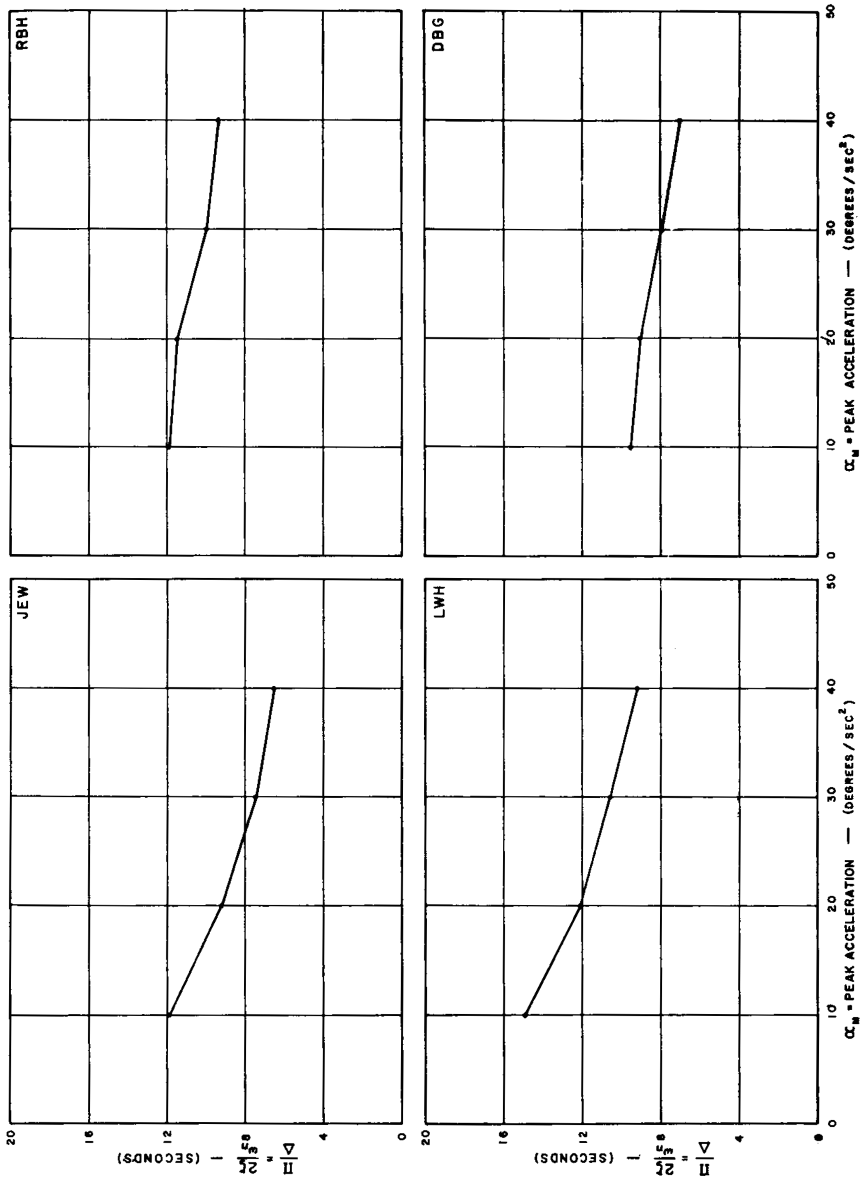
it is possible to calculate the ratios 2 ζ/ω_n or Π/Δ from the phase shift data collected at a known low frequency. (Note that by presenting a single low frequency stimulus to a subject and making this calculation from either his subjective or nystagmus phase shift response, an experimental alternative to the standard cupulometric technique of determining Π/Δ will result.)

By using the nystagmus phase shift data as references these calculations were made for each of the subjects at the 0.02 cps stimulus frequency for each of the four peak acceleration levels and the results plotted as shown in Figure 4. Since, for this stimulus frequency, the phase lag decreases with an increase of peak acceleration level, the ratio of the damping to the stiffness also decreases. This change can then be attributed either to a decrease in the damping of the cupula or to an increase in its stiffness as it is deflected beyond its linear range. Although systems can be devised which become nonlinear due to a decrease of viscous damping with an increase of velocity, it is felt that the velocities of the cupula movements are of such a low magnitude as to make the hypothesis that the stiffness of the cupula increases with increased deflection a more likely interpretation. In fact, the data presented in this report offer qualitative support for the latter viewpoint since, for a given peak acceleration level, any change in stimulus frequency will produce a percentage change in cupula displacement which far exceeds the corresponding percentage change in cupula velocity. A quantitative study in this direction has been initiated.

Further indications that these nonlinearities may be attributed to the magnitude of the cupula deflections can be had by comparing the 0.02 cps data to the 0.04 cps data of Figure 2. Because the 0.02 cps frequency stimulus produces a greater deflection of the cupula than the 0.04 cps stimulus for a given peak acceleration, the former stimulus should produce changes of greater magnitude in the phase shift angle than the latter stimulus; i.e., equal increments of the peak acceleration level should produce greater increments in the phase angle at 0.02 cps than at 0.04 cps. This condition is satisfied by the data of Figure 2 since the slope of the 0.04 cps data is always less than the slope of the 0.02 cps data.

As previously pointed out, the determination of the region of nonlinearity must also be accompanied by an evaluation of the effect of operations in this mode on both the long and short term characteristics of the semicircular canal. If the level and time course of the stimuli used during a single test of this experiment were such as to cause permanent change in the structure of the canal, then these changes would appear in a repeat test. However, the trend of the data indicates that there were no marked effects observable in the present experiment. It is possible that the use of stronger or more sustained stimulus presentations would evoke significant alterations in response in repeated experiments, but this remains to be demonstrated.

Although it is the primary function of this report to delineate the experimental and theoretical considerations involved in using the nys-



Calculated values for the theoretical ratio of the damping to the stiffness of the cupula endolymph mechanism as derived from nystagmus phase shift data produced by a sinusoidal angular acceleration stimulus of cyclic frequency 0.02 cps for various peak acceleration levels.

FIGURE 4

tagmus transition technique to study oculovestibular nonlinearities rather than to derive from the data of these four subjects a specific set of stimulus accelerations and frequencies which approximates linear operation, it is felt worthwhile to demonstrate the approach which may be used for such a quantification.

In assuming the use of sinusoidal angular acceleration stimuli throughout, the first step would involve oscillating the subject at a frequency which would produce a phase lag of 90 degrees between the nystagmus transition and the onset of the related acceleration cycle to obtain the value for the undamped characteristic frequency of the oculovestibular system. From the data of the four subjects used in this study as shown in Figure 2, it would appear that a peak acceleration level between 40 and 60 deg/sec² would satisfy the linearity criteria at these higher stimulus frequencies.

The next step would be to select a rotational frequency much lower than the value determined for the undamped characteristic frequency and to measure the resultant phase shift as peak acceleration is varied. It would then be possible to calculate $2\zeta/\omega_n = \Pi/\Delta$, as is demonstrated in Figure 4, and thus approximate the stimulus level at which nonlinearity of response (i.e., the peak acceleration level which, if exceeded, results in a significant change in phase shift) is considered to be initiated. The theoretical peak cupula deflection which occurs at this acceleration level can then be calculated from

$$\xi_{max} = \frac{\alpha_m}{\omega_n^2} \left[\frac{1}{\left[\left(1 - \frac{\omega^2}{\omega_n^2}\right)^2 + \left(2\zeta\frac{\omega}{\omega_n}\right)^2 \right]^{1/2}} \right] \qquad (5)$$

as was discussed in a previous report (4).

To illustrate this procedure with a numerical example, consider the data of subject RBH as shown in Figure 3 and allow the extrapolation of the phase shift figures to the point where his undamped characteristic cyclic frequency may be assumed to be in the vicinity of 0.3 cps. From Figure 3 note that his phase shift at 0.02 cps with the 10 and 20 deg/sec² acceleration levels is relatively constant and assume that the calculated value of his $2\zeta/\omega_n$, as shown in Figure 4, is approaching 12. The equation of motion for the cupula-endolymph mechanism of RBH in the form of equation (1b) will then be

$$\ddot{\xi} + 42\,\dot{\xi} + 3.5\,\xi = \alpha_m \sin 2\pi Ft$$

$$2\zeta/\omega_n \cong 12$$

$$\omega_n \cong 1.88$$

If it is then desired to assume that nonlinear operation is in effect at the 20 deg/sec² level of the 0.02 cps stimulus, the peak cupula deflection which occurs during each cyclic response will be about 3 degrees as calculated from equation (5), with $F = 0.02$ cps and $\alpha_m = 20$ deg/sec². Any displacements of the cupula in this region or beyond, regardless of the type or form of stimulus, should then result in nonlinearity of response. If these calculations are repeated for subject JEW, a much lower value for the peak cupula displacement will be found since an acceleration level less than 20 deg/sec^2 at 0.02 cps will be necessary to estimate the onset of nonlinearity or, in fact, even the existence of a linear mode of operation.

Obviously this theoretical value for the cupula displacement at which pronounced nonlinearity of response may be observed for RBH will vary from subject to subject, and will not even be singular in value if other criteria are selected for the definition of the onset of nonlinearity. However, the very advantages of the entire technique to approach quantitatively both the definition of oculovestibular system nonlinearities and the identification of individual differences become even more apparent.

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